

Color Consistency of Zirconium Oxide CEREC Crowns Milled at Different Thicknesses

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Purpose: The purpose of this study was to evaluate the color of full-contour zirconia CEREC restorations milled at different material thicknesses which could aid dental practitioners in planning full-contour ceramic restorations.

Method and materials: Two sample crown preparations were made of an individual biogeneric copy of a left maxillary central incisor on a model scanned by means of Sirona CEREC Omnicam. One sample was prepared at a reduction of 1 mm, while the other was prepared at a 2mm reduction. A total of 60 CEREC Ivoclar ZirCad LT blocks were used to create 10 Samples of monochromic Zirconia crowns milled at 1 mm and 2 mm thicknesses in each of Vita classic shades A1, B2, and C2. The optimal thickness of 1 mm was chosen for the control groups Based on manufacturer recommendations and optimal thickness of 2 mm was chosen for the test groups. Sample crowns were sintered by means of a Sirona SpeedFire oven. No additive coloring or glaze was applied. A *Canon 80D* digital camera equipped with a *Canon MR-14EXII* ring flash and a polarized filter was used to photograph all test and control group specimens. The photographs were developed via digital software, *Photoshop CC 2019*. The CIE L*a*b* color values were measured.

Results: CIE L*a*b* data for all samples was recorded and statistically analyzed using a two sample t-test, *STATA.SE v16*. There was a statistically significant difference ($P < 0.05$) in L* values when test groups (2mm) were compared with the control groups (1mm) for shades A1,

B2, and C2 with $P < 0.00$. ΔE for shades A1, B2 and C2 groups were 3.64, 3.67, and 4.55 respectively which is higher than the clinically acceptable threshold 3.3.

Conclusion: An increase in the thickness of Zirconia from 1mm to 2mm demonstrated a $\Delta E > 3.3$ in all test groups which is detectable by untrained observers. As the thickness of Zirconia increased from 1mm to 2 mm, L^* decreased in all test groups. Vita Classic shade C-2 demonstrated a more dramatic decrease in L^* than shade A-1 and B-2 following the Zirconia thickness change from 1mm to 2mm.

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والله ولي التوفيق

1.0 Introduction

Production of naturally appearing ceramic restorations has been a major objective ever since their introduction (1). The opaque core of ceramo-metal restorations limits both color appearance and translucency (2). All ceramic restorations without a metal substructure allow more light transmission and consequently improved reproduction of the appearance of natural tooth structure (3). Despite the esthetic advantage of glass ceramic restorations, their lack of strength has resulted in a demand for increased durability. High strength zirconia-based restorations combined with CAD/CAM technology has broadened the range of employment of ceramics in restorative dentistry (4). Unfortunately, cohesive failures of veneering porcelain has proved to be a major drawback (5, 6). Fabrication of monolithic zirconia restorations consisting of a single zirconia material without veneering porcelain could be an alternative solution (7).

CAD/CAM technology has made it possible to provide patients with accurate, fracture resistant, full contour CEREC restorations in a single day. The aesthetics of zirconia CEREC restorations depends on multiple factors such as material thickness, color, translucency, anatomic features, and shape. Color may vary when zirconia restorations differ in thickness. Evaluation of color of full-contour zirconia CEREC restorations milled at different material thicknesses could aid dental practitioners in planning full-contour ceramic restorations.

2.0 Literature Review

2.1 Zirconia

Zirconium oxide (ZrO_2) is a highly sintered polycrystalline ceramic dioxide of the transition metal zirconium (Zr) which has been utilized in restorative dentistry for approximately two decades. It has occupied a distinctive place among dental ceramic materials because of its superior mechanical properties including high flexural strength, fracture toughness, and low elastic modulus. In addition, zirconia has low corrosion potential, low cytotoxicity, and offers minimal adhesion of bacteria (8, 9). These unique properties have encouraged significant biomedical research since the 1970's for uses of zirconia in medicine and dentistry, particularly in stress-bearing roles where its strength rivals that of many alloys (8).

Zirconia is polymorphic. Its crystal structures or phases can exist as monolithic (m), tetragonal (t) or cubic (c) depending on temperature and pressure. The most stable monolithic phase is at room temperature. As the temperature rises to about 1170°C , the monolithic phase transforms into the tetragonal phase, accompanied by a volume shrinkage of approximately 4-5%. The tetragonal phase evolves into the cubic phase at about 2370°C , with only minimal additional volumetric changes (10, 11). The addition of dopants like yttrium oxide (Y_2O_3), calcium oxide (CaO) or magnesium oxide (MgO) into the ZrO_2 -lattice, stabilizes the tetragonal and the cubic phases at room temperature as metastable phases (12). They can transform to the monolithic phase under the influence of crack initiation in the ceramic. This tetragonal to monolithic phase transformation is associated with 4-5 % volumetric expansion which results in compressive forces at a crack tip slowing its propagation. This unique phenomenon is termed

"transformation toughening" and contributes to Zirconia's high fracture toughness compared to brittle conventional ceramics (13).

The initial generations of dental zirconia were all yttrium stabilized-tetragonal polycrystalline zirconia consisting of fine grain zirconia with small amounts of Y_2O_3 as dopant. These fully crystalline 3Y-TZP ceramics (IPS e.max ZirCAD LT and MO) were composed as follows:

Table 1: Composition of 3Y-TZP

Component	Content
Zirconium oxide (ZrO_2)	88.0 - 95.5 wt%
Yttrium oxide (Y_2O_3)	> 4.5 - \leq 6.0 wt%
Hafnium oxide (HfO_2)	\leq 5.0 wt%
Aluminum oxide (Al_2O_3)	\leq 1.0 wt%
Other oxides for coloring	\leq 1.0 wt%

3Y-TZP commercially available for the fabrication of dental crowns and fixed partial dentures has been processed either by soft machining of pre-sintered blanks followed by high temperature sintering, or by hard machining of fully sintered blocks.

2.1.1 Soft Machining.

Since its development in 2001(14), direct ceramic machining of pre-sintered 3Y-TZP has become increasingly popular and has now been offered by an increasing number of manufacturers. The die or a wax pattern is initially scanned, followed by a computer designed enlarged restoration (CAD). Finally, a pre-sintered ceramic blank is milled by computer aided machining (CAM). The restoration is then sintered at high temperature. Several variations of this process exist depending on the scanning process and compensation for the considerable sintering shrinkage of 3Y-TZP (~25%). This approach has the advantages of rapid milling, reduced cutting forces, increased tool life, potentially better surface quality, and prevention of moisture absorption by the zirconia blanks eliminating the need for drying the milled zirconia prior to sintering (15, 16).

Typically the 3Y-TZP powder used in the fabrication of zirconia blanks contains a binder that enables cold isostatic pressing. The binder is later eliminated during the pre-sintering phase. It also contains about 2% by weight HfO_2 , classically difficult to separate from ZrO_2 . These powders have only minor variations in chemical composition. The powders consist of spray-dried agglomerates of much smaller crystals that are about 40nm in diameter. The blanks are manufactured by cold isostatic pressing. The mean pore size of the compacted powder is very small and in the order of 20–30nm with a very narrow pore size distribution (14).

Binder elimination during pre-sintering heat treatment has to be carefully controlled. If the temperature increase is too rapid, the elimination of binder and associated burn out products can lead to cracking of the blanks. The pre-sintering temperature of the blanks affects the hardness and machinability. Adequate hardness is needed for the handling of the blanks but if the hardness is too great, it might adversely affect machinability. The pre-sintering heat treatment

temperature also affects the surface roughness of a machined blank. Since higher pre-sintering temperatures lead to rougher surfaces, slower heating rates are preferred. The density of each blank is carefully measured so that the appropriate compensating shrinkage occurs during final sintering. The final density of the pre-sintered blanks is approximately 40% of the theoretical density (6.08 g/cm³). The density gradient within the blanks is less than 0.3% of the theoretical density in all directions (14).

2.1.2 Hard Machining

Pre-sintered Y-TZP blocks are processed in a high pressure inert gas atmosphere at temperatures between 1400 and 1500 °C (17, 18). The result is a very high density exceeding 99% of the theoretical density. The blocks can then be machined using a milling system specially designed to handle the increased hardness and difficult machinability of fully sintered Y-TZP (19, 20).

2.1.3 Colour in Dentistry

Achieving natural optical properties using artificial materials is one of the main challenges in restorative dentistry. Color is undoubtedly one of the major parameters considered by patients when judging the esthetics of a restoration (21).

The Munsell Color Order System and the International Commission on Illumination System (CIE) are two principle systems used to describe color. The Munsell system is based on three color coordinates: value describes lightness, hue describes the nature of the color, and chroma describes color saturation. The CIE system is based on the coordinates L*, a*, b*. The

L* coordinate represents the brightness of an object. The a* coordinate represents the red (positive value) or green (negative value) chromacity. The b* coordinate represents the yellow (positive value) or blue (negative value) chromacity (22-24). Among these color parameters, it is generally accepted that the value (L) is the most critical for shade matching. It has been reported that a ΔL of $[\pm 2]$ is the clinically acceptable change threshold (25). Changes related to the a* and/or b* coordinates are better tolerated when clinically assessing color match (26).

A 50:50% perceptibility threshold refers to a situation where 50% of observers notice a difference in colour between two objects while the other 50% perceive no difference. The difference in colour that is acceptable for 50% of observers corresponds to a 50:50% acceptability threshold (AT). If 50% of observers consider a dental restoration to require colour correction while the other 50% consider the colour difference to be acceptable, the difference between those two thresholds is considered the industry tolerance limit and indicates how much perceptible difference can be tolerated while still considering a colour match to be acceptable (27).

It is widely agreed that $\Delta E > 1$ is perceptible (22, 25, 28-49). The acceptability threshold in the literature ranges from ΔE 2.0 to 4.0. The majority of the studies have determined that the 50% acceptability threshold is $\Delta E = 3.7$ (27, 30, 32-35, 41, 43, 44, 46, 48, 50-52). One-third of clinical studies reporting $\Delta E = 3.7$ as a 50% acceptability threshold in the literature refer to the clinical study by Johnston and Kao(30) in 1989. The systematic review stated that recent dental literature is lacking and most of the recent clinical studies refer to studies that have been done three decades ago where the aesthetic demands have been changed.

Color formulas are designed to provide a quantitative representation of color differences between two objects. The most extensively used ΔE formula is derived from the CIE L*a*b*

system, which approximates uniformed distances between color coordinates:

$$\Delta E_{ab} = \sqrt{[(\Delta L)^2 + (\Delta a)^2 + (\Delta b)^2]}$$

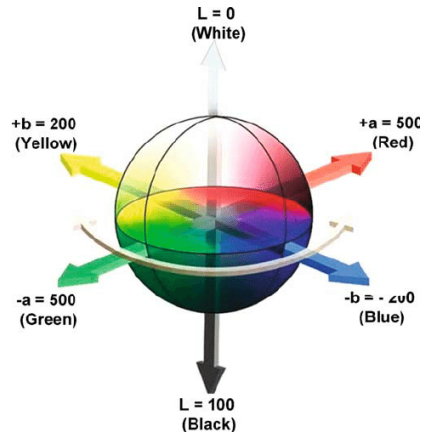


Figure 1: The cubical CIE Lab color space(53)

2.1.4 REVIEW OF COMPUTER-AIDED DESIGN/COMPUTER- AIDED MANUFACTURING (CAD/CAM) SYSTEMS

CAD/CAM fabrication along with the development of new ceramic systems has been replacing conventional lost wax restoration fabrication in restorative dentistry. (54) Duret et al. introduced the commercial Sopha system, in the early 1970's. However, it did not gain popularity due to limitations of the computer systems of that time (54, 55).

By the mid 1980's, the chairside CEREC system was developed by Mormann and colleagues for fabrication of ceramic inlays and onlays (56).

In 1987, Swiss dentist, Dr. Werner Mörmann, and Italian electrical engineer, Marco Brandestini, introduced the first digital intraoral scanner which evolved into CEREC® by Sirona Dental Systems LLC (Charlotte, NC) which was the first commercially available CAD/CAM system for dental restorations (57, 58). Since then many different digital impression and CAD/CAM milling systems have been introduced. With the availability of systems capable of capturing 3D virtual images from the tooth preparation, chairside restorations can be made either directly via CAD/CAM systems or remotely at a dental laboratory from an accurate master model of the tooth preparation(57).

3.0 Null Hypothesis

. There is no change in color consistency of CEREC Zirconia crowns milled at 1 mm and 2 mm thicknesses using Vita Classic shades A 1, B 2 and C2

4.0 Method and Materials

Using *Sirona CEREC Omnicam*, a dental model was scanned and an individual biogeneric copy of the maxillary left central incisor was generated (figure 2). Two sample crown preparations were made using depth cut and diamond chamfer burs. One sample was prepared with a 1 mm reduction while the second sample was prepared with a 2 mm reduction. The crown preparations were scanned by means of a *Sirona CEREC Omnicam*. Preparation thicknesses were verified by superimposing the biogeneric individual copy and utilizing preparation analysis tools (Figure 3, 4, 5). Ivoclar ZirCad LT was used as the fabrication material of choice. A total of 60 CEREC MC XL blocks were used to create 10 Samples of monochromic Zirconia crowns milled at 1 mm and 2 mm thicknesses in each of Vita classic shades A 1, B 2, and C 2 (Figure 4, 5, 6). In addition, all sample blocks had the same LOT number. Crowns were sintered by means of a SpeedFire oven (Figure 7), *Sirona*. No additive coloring or glaze was applied. Each sample crown was inserted into the prepared dental models with the identical stump shade.



Figure 2: Biogeneric Individual copy of tooth #9



Figure 3: Biogeneric Copy Supperimposed with Crown Preparation



Figure 4: Verifying 1 mm Middle Thrid Crown Thickness



Figure 5: Verifying 2 mm Middle Thrid Crown Thickness

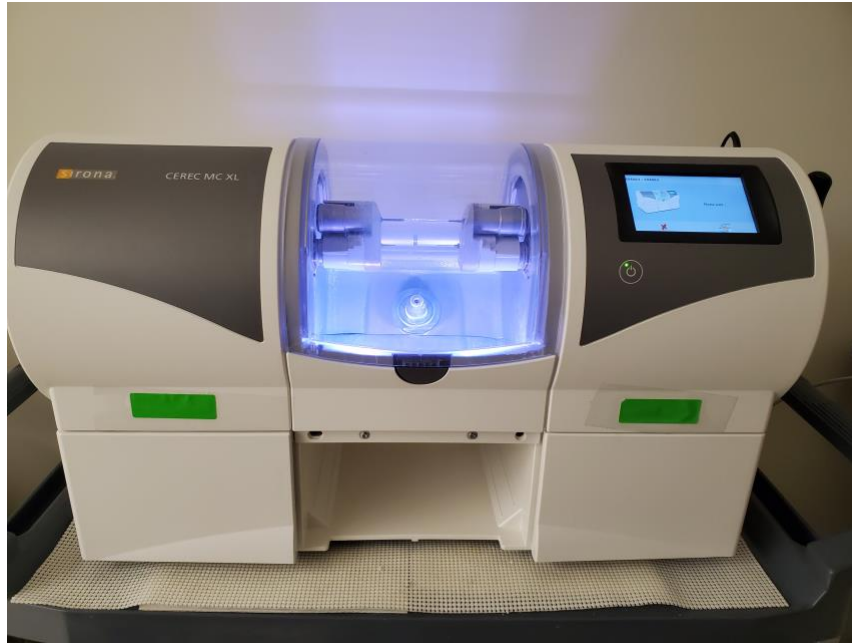


Figure 6: CEREC MC XL Milling Machine

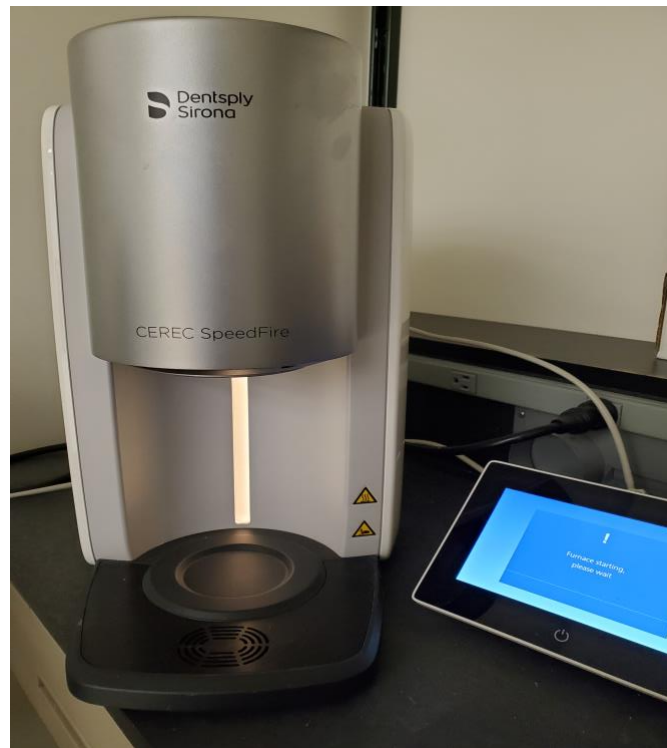


Figure 7: Crown Sintered in Sirona Speed Fire Furnace

4.1 Control and Test Groups

Based on manufacturer recommendations for Ivoclar ZirCad LT, the optimal thickness of 1 mm was chosen for the control groups for shades A 1, B 2, and C 2. The optimal thickness of 2 mm was chosen for the ZirCad crowns of the test groups for shades A 1, B 2, and C 2.

Table 2: Control and test groups of sample

Group	A 1	B 2	C 2
Control (1 mm)	10	10	10
Test (2 mm)	10	10	10

4.2 Color Measurements

A *Canon 80D* digital camera equipped with a *Canon MR-14EXII* ring flash was used to photograph all test and control group specimens. All the photographs were exposed using the same camera settings: ISO 100, Shutter Speed 1/125, F 22. The camera was mounted on a tripod to control object to lens distance (Figure 9). A polarized filter, *Polar_eyes*, was employed to remove any flash glare as (Figure 10 and 11). Photographs were taken one minute apart to allow the ring flash to fully recharge. The photographs were developed via digital software, *Photoshop CC 2019*. The CIE L*a*b* color values were measured as shown in Figure 12.

The CIE color space of each reading was measured and recorded in terms of the 3 coordinate values (L*, a*, b*). Mean coordinate values and the standard deviations (SD) were calculated for each group by means of two-sample t-test. ΔE (color difference value) was calculated between the control group and test groups' means, according to the formula:

$$\Delta E_{ab} = \sqrt{[(\Delta L^*)^2 + (\Delta a^*)^2 + (\Delta b^*)^2]}$$



Figure 8: Polar_eyes Cross Polarization Filter



Figure 9: Camera Mounted on Tripod with a Fixed Lens-to-Object Distance

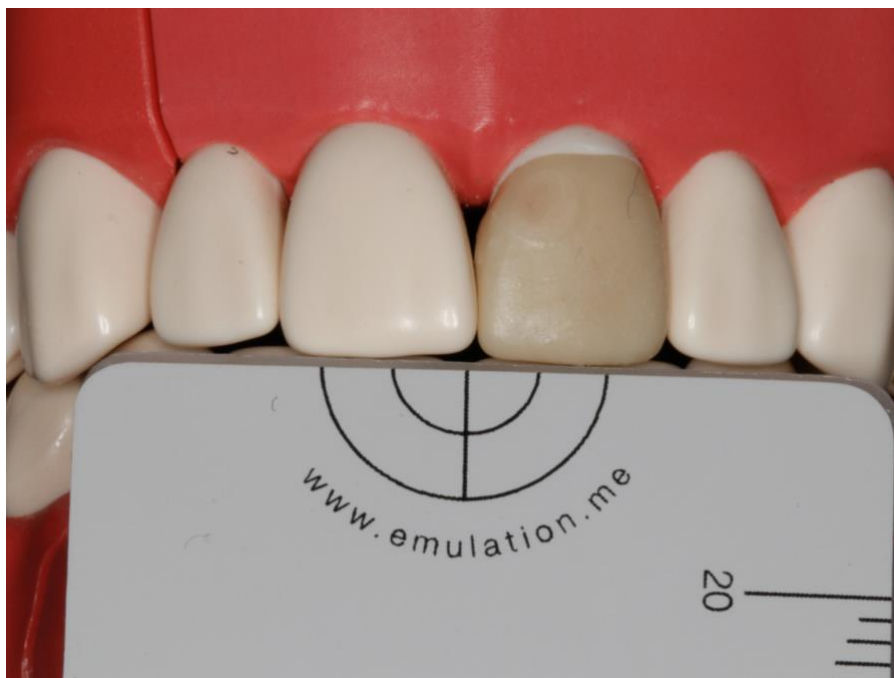


Figure 10: Photo Taken Without Polirization Showing Light Reflection

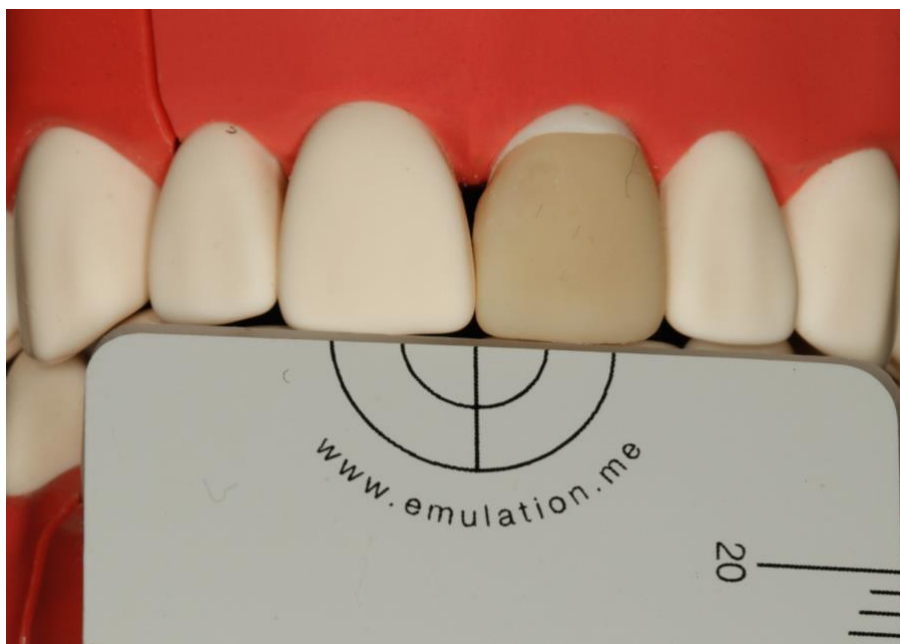


Figure 11: Photo Taken With Polirization to Remove Glare and Light Reflection



Figure 12: : CIE L*a*b* Analysis Using Adobe Lightroom CC

5.0 Results

CIE L*a*b* data for all samples was recorded and statistically analyzed using a two-sample t-test, *STATASE v16*. ΔE mean values and standard deviations of the test and control groups are listed in (table 3). There was a statistically significant difference ($P < 0.05$) in L* values when test groups (2mm) were compared with the control groups (1mm) for shades A1, B2, and C2 with $P < 0.00$. The a* values showed a statistically significant difference when the test groups were compared with control groups for shades A1 and B2 with $P < 0.00$. However, for shade C2 there was no statistically significant difference ($P = 0.6$). The b* values showed a statistically significant difference when test groups were compared with the control groups for shade C2 with $P < 0.00$ while for shades A1 and B2 there was no statistically significant difference with $P = 0.44$ and 0.55 respectively. The ΔE values for all test groups were above the perceptibility threshold ($\Delta E > 1$). ΔE for shades A1 and B2 groups were 3.64 and 3.67 respectively which is more than the clinically acceptable range ($(\Delta E) < 3.3$), while the ΔE for C2 group was 4.55 which is significantly higher than the clinically acceptable range.

**Table 3: Mean CIE L*a*b* values (standard deviation) of test group with their color differences (ΔE)
compared to the control group**

Control and Test Groups	L* (SD)	a* (SD)	b* (SD)	ΔE
A1 (Control) 1 mm Thickness	72.86 (0.17)	2.27 (0.42)	11.97 (0.66)	
A1 (Test) 2 mm Thickness	69.32 (0.23)	3.09 (0.34)	12.21 (0.70)	3.64
B2 (Control) 1 mm Thickness	69 (0.19)	3.64 (0.47)	17.24 (1.34)	
B2 (Test) 2 mm Thickness	65.47 (0.12)	4.58 (0.70)	17.59 (1.22)	3.67
C2 (Control) 1 mm Thicknes	63.05 (0.15)	6.17 (0.88)	20.9 (0.86)	
C2 (Test) 2 mm Thickness	59.16 (0.15)	6.16 (0.59)	18.54 (0.99)	4.55

6.0 Discussion

This study evaluated the effect that different ceramic thicknesses had on the final color of monolithic zirconia crowns. The results revealed a significant difference in CIE Lab and ΔE values related to the zirconia thickness. Therefore the null hypothesis was rejected.

Different values of ΔE in terms of perceptibility and acceptability have been reported in the literature. Vichi et al.(59) divided ΔE into three ranges where ΔE less than 1 is undetectable by the human eye, ΔE values greater than 1 but less than 3.3 though detectable by a skilled operator are considered clinically acceptable, and ΔE values greater than 3.3 are observable by an untrained observer and are considered unacceptable (60-63). Accordingly, this study considered $\Delta E = 3.3$ as the acceptability threshold to evaluate color difference among the test samples. The increased thickness of zirconia from 1 mm to 2 mm demonstrated a color difference of $\Delta E > 3.3$ in all test groups which was detectable by untrained observers.

This study found that as the thickness of zirconia increased from 1 mm to 2 mm, L^* values decreased in all test groups. The thickness of the zirconia not only affected the color, but the selected shade of test crowns. Of note in this study was the fact that shade C 2 had a more dramatic color change ($\Delta E = 4.55$) compared to shades A 1 and B 2 (ΔE 3.64 and 3.67 respectively) when the thicknesses of the zirconia was increased. Consequently, as the value (brightness) of a selected shade decreased (from shade A 1 to C 2), increasing the thickness of a zirconia crown may exhibit an increased ΔE color change.

Tabatabaian et al (64). tested shade A-2 monolithic zirconia specimens with thicknesses of 0.7, 0.9 and 1.1 mm from 2 different manufacturers. They found that as the zirconia thickness increased the L^* value decreased and the impact of the ΔE change on the final color was

significant regardless of the brand of zirconia. The ideal thickness of a zirconia restoration should be 0.9 mm in order to match the targeted shade. The results of this study were similar. The zirconia crowns used in the control groups had a thickness of 1 mm. The zirconia specimens in the test groups had a thickness of 2 mm and showed a decrease in L^* values along with significant changes in ΔE .

Kim et al. (65) studied the color change of 2 mm thickness zirconia specimens after being reduced 0.1 mm at a time until 1 mm thickness is reached. The study showed a noticeable color change $\Delta E > 3.7$ even after the first 0.1 mm of reduction. They also observed that L^* values decreased as the thickness of zirconia crowns increased. This can be explained by the increased absorption of light by the thicker specimens. The authors stated that using only shade A 2 was a significant limitation of their study.

Giti and Hojati (66) found that for zirconia specimens in shade A 2, a decrease in thickness from 2 to 1 mm resulted in a clinically detectable color difference ($\Delta E > 3.7$) as well as an increase in the L^* values of the specimens. The authors stated that using only shade A 2 was a limitation of their study and suggested that further research was needed using different shades of zirconia specimens.

7.0 Conclusion

1. An increase in the thickness of Zirconia from 1mm to 2mm demonstrated a $\Delta E > 3.3$ in all test groups which is detectable by untrained observers.
2. As the thickness of Zirconia increased from 1mm to 2 mm, L^* decreased in all test groups.
3. Vita Classic shade C-2 demonstrated a more dramatic decrease in L^* than shade A-1 and B-2 following the Zirconia thickness change from 1mm to 2mm.

Appendix A

Appendix Table 1: Shadde A 1 CIE L*a*b* Data

1 mm Thickness			2 mm Thickness		
L*	a*	b*	L*	a*	b*
72.9	2.5	11.8	69.2	2.9	12
73	1.9	12.4	69.1	2.5	12.7
72.8	2.9	13	69.4	2.8	11.8
72.7	2	13.2	69.7	2.7	12.6
72.8	2.3	11.4	69.6	3.4	12
72.5	2.5	11.5	69.4	3.4	11.5
72.9	2.7	11.4	69.2	3.3	12.6
73.1	2.3	11.7	69	3.1	13.7
73	2.2	11.7	69.1	3.5	11.9
72.9	1.4	11.6	69.5	3.3	11.3

Appendix Table 2: Shade B 2 CIE L*a*b* Data

1 mm Thickness			2 mm Thickness		
L*	a*	b*	L*	a*	b*
68.8	3.6	17.8	65.5	4.8	18
69.1	4.4	16.1	65.4	5.6	15.5
68.9	3.5	19.3	65.2	4.2	19.6
69.1	3.4	16.3	65.5	3.6	18.6
69.1	3.4	19.3	65.5	5.3	17.7
68.8	4.1	16.6	65.6	4.9	18.4
68.7	3.6	16.6	65.6	4.3	16.1
69	2.7	16.6	65.6	5.3	16.7
69.2	4.1	18.3	65.4	3.7	17.3
69.3	3.6	15.5	65.4	4.1	18

Appendix Table 3: Shade C 2 CIE L*a*b* Data

1 mm Thickness			2 mm Thickness		
L*	a*	b*	L*	a*	b*
63.2	6.5	21.9	59.1	5.7	18.8
63.1	6.5	20.6	59.1	6.5	18.8
63	6.8	20.3	59.4	5.3	19
63.1	6.7	19.6	58.9	6.8	17.4
63.1	6.8	20.5	59.3	6.2	16.7
63.3	7.2	20.6	59.3	5.3	18.7
62.9	5.6	20.8	59	5.9	18.7
62.8	4.2	20.5	59.1	6.9	17.7
63.1	5.8	21.7	59.2	6.3	19.8
62.9	5.6	22.5	59.2	6.7	19.8

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